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ORIGINAL ARTICLE

Mechanical Load on the Upper Extremity During Wheelchair Activities

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ABSTRACT. van Drongelen S, van der Woude LH, Janssen TW, Angenot EL, Chadwick EK, Veeger DH. Mechanical load on the upper extremity during wheelchair activities. Arch Phys Med Rehabil 2005;86:1214-20.

Objective: To determine the net moments on the glenohumeral joint and elbow joint during wheelchair activities.

Design: Kinematics and external forces were measured during wheelchair activities of daily living (level propulsion, riding on a slope, weight-relief lifting, reaching, negotiating a curb) and processed in an inverse dynamics biomechanic model.

Setting: Biomechanics laboratory.

Participants: Five able-bodied subjects, 8 subjects with paraplegia, and 4 subjects with tetraplegia.

Interventions: Not applicable.

Main Outcome Measure: Net moments on the glenohumeral joint and elbow joint.

Results: Peak shoulder and elbow moments were significantly higher for negotiating a curb and weight-relief lifting than for reaching, level propulsion, and riding on a slope. Overall, the elbow extension moments were significantly lower for subjects with tetraplegia than for those with paraplegia.

Conclusions: The net moments during weight-relief lifting and negotiating a curb were high when compared with wheelchair propulsion tasks. Taking the effect of frequency and duration into account, these loads might imply a considerable risk for joint damage in the long term.

Key Words: Activities of daily living; Biomechanics; Elbow; Rehabilitation; Shoulder; Wheelchairs.

IN HANDRIM WHEELCHAIR USERS, the upper extremities are at serious risk of overuse injuries. Wheelchair use requires continuous use of the upper extremities, not only for mobility but also for transfers, weight-relief lifts, and reaching activities. Studies[1,2] have shown that shoulder pain and impingement frequently occur among people with a spinal cord injury (SCI). Pain is experienced during wheelchair-related activities of daily living (ADLs), such as wheelchair propulsion and performing transfers. Because these activities are essential for functional independence, quality of life, and even the life expectancy of people after an SCI,[3] evaluating the mechanical load on the shoulder is important to an understanding of the mechanisms that may cause upper-extremity joint degeneration. Factors that have been mentioned as contributors to the development of shoulder complaints are the relatively high load and high frequency of this load on the shoulder during wheelchair propulsion.[4] In addition, and possibly even more important, the load on the shoulder during other wheelchair-related tasks, such as transfers and weight-relief lifts, has been mentioned.[5-7]

In our study, we used net moments around the elbow and the glenohumeral joint (GHJ) to quantify the mechanical load on those joints. Net joint moments are generally used to analyze (working) conditions and to classify these conditions.[8] To show the high loading at the shoulder, studies[9-12] have presented net joint moments for wheelchair propulsion at various speeds and for varying external power outputs.

Some studies have reported high net moments during ADLs and work-related activities of able-bodied subjects[13-15]; however, little is known about the mechanical load during wheelchair-related ADLs.

In studies with able-bodied subjects, Anglin and Wyss[13] reported unilateral net moments on the shoulder of 16Nm for coming from sit to stand and 28Nm for lifting a suitcase; Kuijer et al[14] calculated net moments between 10 and 30Nm for pulling a refuse container.

Harvey and Crosbie[7] are the only authors thus far who have estimated shoulder and elbow moments (respectively, 45Nm and 30Nm) for subjects with tetraplegia during a weight-relief maneuver. Muscle activity was studied by Reyes,[5] Perry,[6] and Newsam[16] and colleagues, who showed high muscle activation of the latissimus dorsi, the long head of the triceps, and the sternal part of the pectoralis major during transfers and weight-relief maneuvers, respectively. The study by Harvey and Crosbie[7] reported far higher shoulder moments than were found for ADL wheelchair propulsion.[10] It is likely that wheelchair-related daily activities can result in higher peak mechanical loads on the shoulder (especially) than everyday wheelchair propulsion. However, until now, no systematic analysis of several wheelchair-related ADLs for both able-bodied and SCI subjects has been conducted.

The aim of this study was to compare the mechanical load between subjects with a high-level SCI to subjects with a low-level SCI. Subjects with a high-level SCI show a higher prevalence and intensity of shoulder pain than subjects with a low-level SCI.[2] Not only are key muscles, such as the triceps brachii, lattisimus dorsi, and the sternal part of the pectoralis major, often compromised,[17,18] but subjects with a high-level SCI also have less trunk control. It is to be expected that more compensatory activity is needed in the remaining shoulder

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Table 1: Subject Characteristics

Characteristics	Able-Bodied (n=5)	Paraplegic (n=8)	Tetraplegic (n=4)
Age (y)	22±3	39±12*	28±5
Height (m)	1.82±0.11	1.86±0.08	1.88±0.05
Weight (kg)	73±5	79±9	70±14
Injury level	NA	T3-12	C6-7
Years after SCI	NA	14±10	7±6

NOTE. Values are mean ± standard deviation (SD) or range.
Abbreviation: NA, not applicable.
*Significantly different compared with able-bodied ($P<.05$).

muscles to stabilize the GHJ, which might be revealed by kinetic and kinematic analysis. Combined with different kinematics, this will be reflected in a difference in, and likely higher, net moments for the shoulders.

The aims of this study were to determine (1) the net moments acting on the shoulders and the elbows during various wheelchair-related activities and (2) the differences between net moments on the GHJ and elbow joint for subjects with a high-level or low-level SCI versus able-bodied subjects.

METHODS

Participants

Seventeen subjects participated (table 1): 5 able-bodied subjects, 4 with tetraplegia, and 8 with paraplegia. Two subjects with paraplegia and 1 subject with tetraplegia had an incomplete lesion. The inclusion criteria for this study were that subjects be male and have no current history of shoulder problems. All subjects were informed about the nature of the study before giving written informed consent to participate. The protocol of this study was approved by the Medical Ethical Committee of the Vrije Universiteit Medical Center.

Protocol and Tasks

To determine the net moments during wheelchair-related ADLs, subjects performed different standardized ADLs under experimental conditions in an instrumented wheelchair (fig 1). Both 3-dimensional external forces and moments and 3-dimensional kinematics of the upper extremity were determined in each activity. Before testing, all subjects were allowed to become accustomed to the experimental wheelchair and the experimental setup.

Subjects performed 3 tasks: wheelchair propulsion, a weight-relief lift, and a reaching task. The subjects with an SCI performed 2 additional tasks. Wheelchair propulsion was performed at .83ms, to ensure a submaximal exercise level for all subjects. When the level treadmill[a] was at speed and the subject was propelling comfortably, data were collected for a period of 30 seconds.

Because of the design of the recording system, the weight-relief lift had to be performed with the hands on the handrims. However, subjects were allowed to place the left (nonmeasured) hand on the tire, to create a larger support base. This task was performed 3 times with 20-second rests between trials.

The third task was placing different bottles on a platform, 0.5m off the ground. The bottles varied in mass (0.1, 0.75, 1.5kg). At the start of each trial, subjects sat in the wheelchair and held the bottle at their lap; subsequently, they placed the bottle on the platform in front of them and took it back to the starting position. For this task, the exerted hand force was the force needed to compensate for the gravitational force on the bottle.

Subjects with SCI also performed the following 2 additional tasks: subjects had to propel on a slope of 3% at a speed of .56ms. When the treadmill reached the preset slope, 30 seconds of propulsion were recorded.

The fifth task was negotiating a curb of 10cm. Before negotiating the curb, the subjects were allowed to practice with the experimenter behind the wheelchair. If the subject was not comfortable performing the task, the task was cancelled. If the subject was comfortable, 3 successful trials were recorded.

Instrumented Wheelchair

All tasks were performed in a Quickie Triumph[b] wheelchair (see fig 1). A 6 degrees of freedom AMTI force transducer[c] was built into the right wheel. The handrim was connected to the transducer by an aluminum shell. Next to the transducer, a portable data acquisition device[d] and an angular position sensor[e] were built into the wheel.

The wheelchair had a standard design with the backrest of the chair .42m wide and .40m high. The seat was .42m wide and deep. Seat height was .55m, seat angle to the horizontal

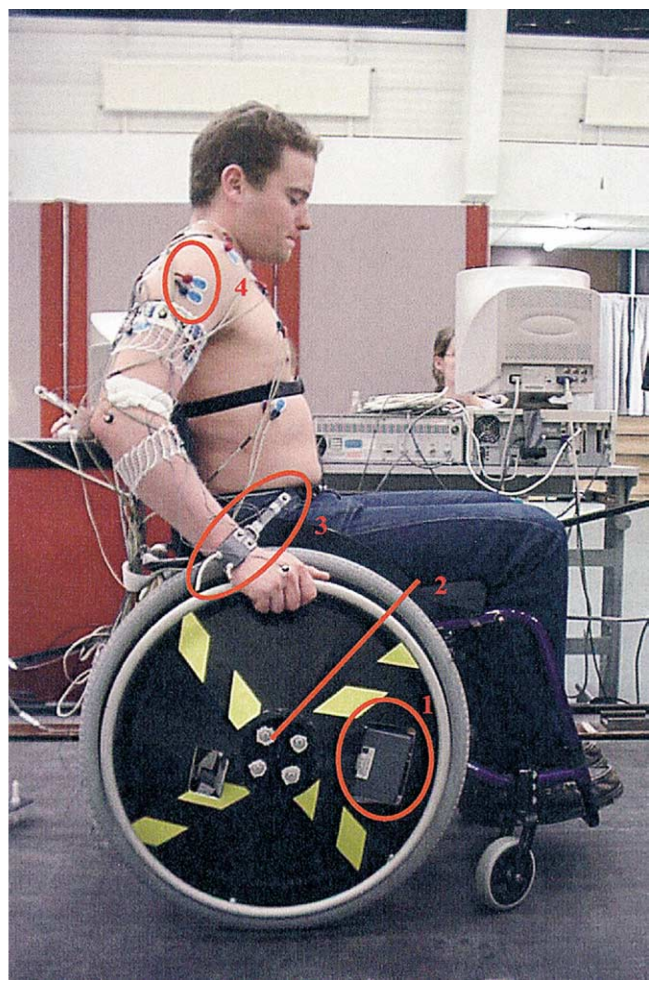

Fig 1. Able-bodied subject in instrumented wheelchair during level wheelchair propulsion. Legend: 1, data logger; 2, instrumented wheel; 3, technical marker; 4, electromyographic electrodes (electromyography not used in our study).

was 10°, and the angle of the back to the vertical was 5°. The radius of the wheels and rims were, respectively, .305 and .265m. The diameter of the rim tube was 20mm, the pressure of the rear tires was 4.5 bar, and the camber of the wheels was set at 5°. After the instrumented wheel was balanced, the inertia was calculated; subsequently, the inertia of the other wheel was corrected by adding extra weights. The total weight of the instrumented wheelchair was 18.6kg.

Data were stored on a memory Flash card. The instrumented wheel enabled us to measure the (propulsive) forces applied on the handrim as well as the torques on the handrim. The hand torque applied by the hand on the rim was calculated from the difference between the torque that was measured around the wheel axis and the torque produced by the applied force on the handrim.[9] It was assumed that the force was applied at the third metacarpal as the point of hand contact. The accuracy of the instrumented wheel was measured in newtons (Fx = forward = 3.0N, Fy = downward = 2.8N, Fz = medial = 4.1N) and for the moments in newton meters (Mx=0.3Nm, My=0.7Nm, Mz=0.4Nm).

The AMTI force transducer was synchronized with the Optotrak computer[f] by a telemetric system.[e] Forces and torques were low-pass filtered by using a 10-Hz second-order recursive Butterworth filter. All torques and forces from the wheelchair were transformed from the rotating (local) coordinate system of the force transducer to forces and torques in the global coordinate system and subsequently corrected for the camber of the wheelchair and for the offset; the weight of the rim and the shell connected to the transducer.

Kinematics

Kinematics were recorded with a 3-camera optoelectronic system.[f] Seventeen active markers were placed on the right side of the subject's body (thorax, upper arm, forearm, hand) as well as on the wheelchair.[19,20] The 3-dimensional positions of markers were recorded at 100Hz during each experimental trial. Recordings were performed with technical markers on the epicondylus medialis humeri and the processus styloideus ulnae. Before the actual measurements, a calibration measurement was performed in which the orientation of the technical markers was defined relative to bony landmarks. Also, the orientation of the scapula was determined by a calibration measurement with a scapula-locator system,[21] while the subject sat in the wheelchair with the arm in the anatomic position. From the scapula calibration measurement and the orientation of the humerus during the tasks, the orientation of the scapula and clavicula were calculated by using a regression model of Pascoal.[22] From the position of the landmarks the local coordinate systems of the trunk, humerus, and forearm were reconstructed according to the guidelines of the International Shoulder Group.[20]

Biomechanic Model

The kinematics of the right arm and shoulder and the exerted forces at the hand were used as input for the Delft Shoulder and Elbow Model.[23,24] The input kinematics derived from the position of the incisura jugularis and the orientation of the thorax, humerus, forearm, and wrist. Orientation of the scapula and clavicula was obtained from regression equations. Further, the 3-dimensional external forces and the torques applied by the hand on the rim were used as input. Output variables of the model used in this study were net joint moments around the GHJ and around the elbow joint.

Data Analysis

The moments around the GHJ were expressed as moment components (flexion and extension, endo- and exorotation, abduction and adduction) relative to the thorax. The moment components were used to calculate the resultant net moments on the GHJ. The net elbow moment was calculated as the moment around the flexion-extension axis of the elbow joint only (extension = positive, flexion = negative).

From the 30 seconds recorded during the wheelchair propulsion tasks, 5 consecutive pushes were selected for data analysis. For every push, the peak net shoulder and elbow moment were determined. The push phase was defined as the phase in which the external force was above the level of noise in the recovery phase. For the other ADL tasks the peak values for each trial were determined. However, to compare the reaching task with the other tasks, only the peak shoulder and elbow moments of the trial with the 1.5-kg weight were used. This specific trial was chosen to create a broader range of variation of external loading and thus of net moments. For the weight-relief lift, the moments around the elbow and the shoulder were corrected for body weight because the applied forces highly depend on the body mass.

Statistical Analysis

To detect significant differences among the subject characteristics of the 3 subject groups, independent *t* tests were applied.

For each task, the mean of the peak moments over the trials was calculated. To compare the peak moments among the tasks a general linear model for repeated measures was used (within-subject factor: task; between-subject factor: groups). Depending on the tasks that were compared, different numbers of subjects were used. The level of significance was set at *P* less than .05 for all statistical tests.

RESULTS

Participants

All subjects were able to perform the requested tasks except for negotiating the curb. The latter task could be performed by only 5 of 8 subjects with paraplegia. The data of 1 of the tetraplegia subjects worked out to be erroneous because of missing values in the Optotrak data for both propulsion tasks and had to be discarded.

Except for age between the able-bodied and the paraplegic group, no differences were found for subject characteristics.

Wheelchair Propulsion

The peak net moments for the shoulder and elbow for low-intensity wheelchair propulsion were between 4.1 and 11.3Nm and between −0.5 and 7.9Nm, respectively (table 2). The highest components around the shoulder were the adduction and the anteflexion components. Figure 2 gives a typical example of the net shoulder moment during the whole push.

No significant differences were found between the able-bodied subjects and the subjects with a high or a low SCI for both the shoulder and elbow peak moments.

Weight-Relief Lift

For the weight-relief lift, the peak moments on the shoulder and the elbow for the 3 trials of lifting were calculated. Figure 3 gives a typical example of the moments around the GHJ. The 2 large moment components at the shoulder were retroflexion and adduction. After correction for body mass, the mean peak shoulder

Table 2: Peak Net Shoulder and Elbow Moments for the 5 ADL Tasks for Able-Bodied Subjects and Subjects With Paraplegia and Tetraplegia

Tasks	Moments	Able-Bodied (n=5)	Paraplegic (n=8)	Tetraplegic (n=4)	P
Level wheelchair propulsion	GH peak (Nm)	6.7±2.8	7.2±2.4	9.0±1.8	
	EL peak (Nm)	3.6±2.0	3.0±2.3	2.0±2.3	
	Power output (W)	5.1±0.8	4.4±0.5	4.3±1.6	
Reaching	GH peak (Nm)	12.3±1.8	12.3±1.0	13.6±0.5	
	EL peak (Nm)	−5.5±0.4	−5.6±0.2	−4.6±1.6	
Riding a slope	GH peak (Nm)	NA	14.6±3.8	18.0±1.8	
	EL peak (Nm)	NA	5.7±2.1	7.6±2.5	
	Power output (W)	NA	10.8±2.1	11.8±0.9	
Weight-relief lift	GH peak (Nm)	36.1±8.7	43.9±9.4	44.4±17.9	
	EL peak (Nm)	35.3±10.6	42.2±6.2	21.5±11.9	.008
	GH peak (Nm/kg)	0.5±0.1	0.6±0.1	0.6±0.1	
	EL peak (Nm/kg)	0.5±0.1	0.5±0.1	0.3±0.2	
Negotiating a curb	GH peak (Nm)	NA	75.1±23.5	NA	
	EL peak (Nm)	NA	60.2±16.5	NA	

NOTE. Values are mean ± SD.
Abbreviations: EL, elbow joint; GH, glenohumeral joint.

moment was .56Nm/kg and for the elbow .47Nm/kg for all subjects.

The absolute peak net moments for the shoulder and elbow were, respectively, between 24 and 70Nm and between 8 and 51Nm (see table 2). The absolute values were used to compare the weight-relief lift with the other ADL tasks. For the shoulder moments, no significant differences were found between the subject groups.

However, a significant difference was found for the absolute peak elbow moment between the subject groups ($P=.008$), leaving only a trend for the elbow moments when corrected for body mass ($P=.062$). For comparison with the other tasks, the average over the 3 trials of the peak moments was calculated.

Reaching

The peak shoulder and elbow moments for the 3 trials of placing a bottle on the shelf were calculated (fig 4, see table 2). The net moment on the GHJ increased from 5.8 to 12.7Nm with the increasing weight, the net moment on the elbow increased from –1.1 to –5.2Nm. A flexion moment in the elbow was needed to hold up the weight, whereas a mainly anteflexion moment in the shoulder was needed to hold up the arm in front of the body. For a higher weight, significantly higher shoulder and elbow moments were found ($P<.001$). No significant differences were found between the 3 groups.

Propelling on a Slope

Only the subjects with SCI performed this task. The peak net moments for the shoulder were between 9.7 and 20.6Nm and for the elbow between 3.3 and 9.7Nm (see table 2). No significant difference was found for the net moments between the subjects with paraplegia and with tetraplegia.

Negotiating a Curb

Five of 12 subjects with an SCI were able to perform this task in the experimental wheelchair. The peak shoulder mo-

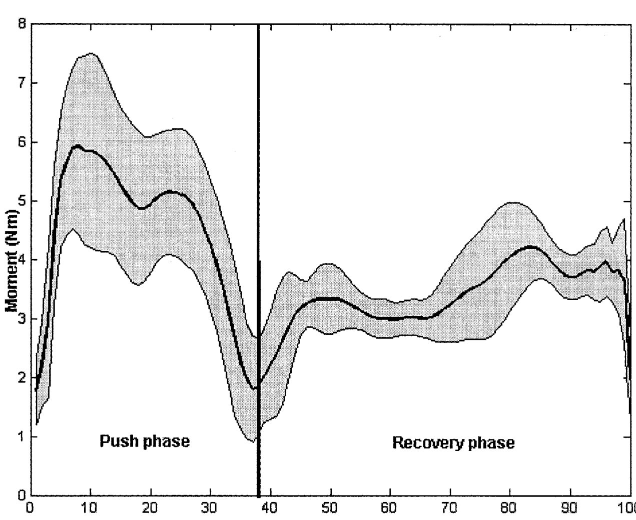

Fig 2. Typical example of the net shoulder moment (paraplegia subject) during wheelchair propulsion. Mean over 5 pushes and standard deviation (SD), time normalized to a full cycle (100%).

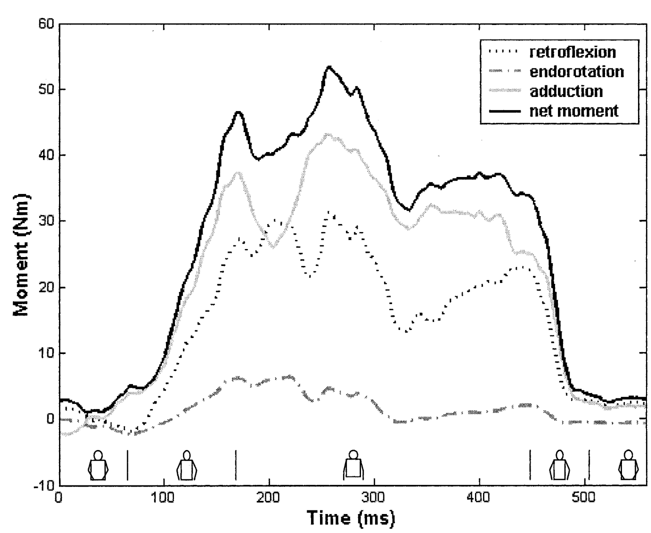

Fig 3. Typical example of the net shoulder moment and moment components for an able-bodied subject during a weight-relief lift.

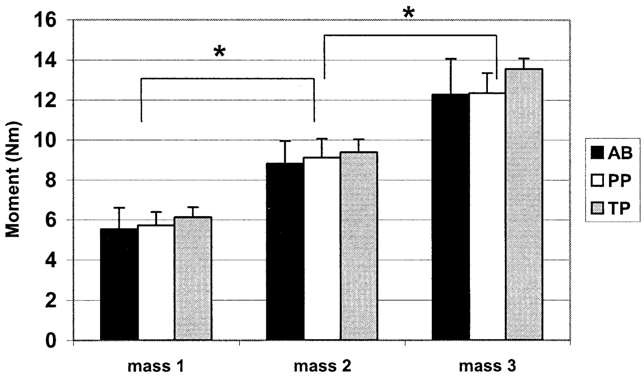

Fig 4. Net shoulder moments (mean, SD) during the reaching tasks for able-bodied subjects (AB) and subjects with paraplegia (PP) and tetraplegia (TP) for 3 different mass conditions. Reaching height was 0.5m. *Significantly different (*P*<.001).

ments were between 36 and 97Nm for the different subjects (see table 2). The elbow moments were between 32 and 75Nm.

DISCUSSION

This study was conducted to gain insight into the external loading on the shoulder and the elbow during various wheelchair-related ADLs among subjects with SCI and nonimpaired subjects. Although the mechanical load of wheelchair propulsion has been studied extensively, few studies[5-7,16] have looked at wheelchair-related ADLs.

Wheelchair Propulsion Tasks

Subjects propelled the wheelchair on a level surface at a speed of .83m/s as well as at a speed of .56m/s on a slope of 3%. Therefore, external power output was limited; that is, 4.6±0.9W for level propulsion and 11.0±1.9W for riding on a slope. However, a setup with a low speed without extra resistance was chosen, so that all subjects were able to ride at a submaximal level. The net moments we found seem to deviate from other studies[9-11,25,26] on wheelchair propulsion; however, considering the differences in power output among these studies and our study, the net moment values we found did not differ from those in the literature.

Weight-Relief Lift

Apparently, subjects with a high lesion level performed this task in a somewhat different way because they were not able to use full triceps activity to extend their arms. They seemed to first lock the elbow joint, after which they lifted their body weight from the shoulder with the clavicular part of the pectoralis muscle and the deltoid muscle. Therefore, the trend for a difference in elbow moments between the subjects with a high and low SCI could be explained by different kinematics. However, different activation levels of the triceps can cause these differences as well.

In addition, the constraint that subjects were required to use the handrim (and the combination of handrim and tire on the left side) to lift themselves may have influenced the position and orientation of trunk and arms and thus could have had an influence on the direction of the exerted forces and the magnitude of the net moments. For small subjects, the handrims were further away from the body center than for larger subjects, which may increase external loading further. The elbow extension during weight-relief lifting (able-bodied, 18°±4°; paraplegia, 29°±6°; tetraplegia, 20°±11°) is a risk factor, and in combination with the high elbow moment it could compromise the integrity of the elbow joint.

Recently, the guidelines for pressure relief by weight-relief lifting have been revised in the Netherlands,[27] as a result of a study by Coggrave and Rose,[28] who found that traditional lifting was not efficient. In the light of the results of our study and other research[13,28] this policy change makes sense.

Reaching

The results of the reach tasks showed that the net moments are dependent on the mass of the object. However, the actual moments were probably underestimated because of the low-segment mass in the model. The moment on the shoulder joint during reaching with the empty bottle was comparable to wheelchair propulsion at a low speed. Because reaching with a weight of 1.5kg is a much more straining task, the shoulder moments were almost identical to the moments for propelling on the slope.

In this study, all subjects were able to perform all the tasks, and no differences were found between groups. This implied that no essential differences were found in kinematics and in external forces. For subjects with a high lesion level, different strategies may be necessary to stabilize the joint as a result of partial muscle paralysis. This compensation activity may lead to a high muscle stress and/or a high joint reaction force.

Negotiating a Curb

Only 5 subjects with a low-level SCI (of our 12 subjects with SCI) could negotiate the curb. These subjects were very fit and were well able to handle the rather heavy (18.6kg) experimental wheelchair. Clearly, this task is accompanied by very high net moments in both the shoulder and the elbow. Subjects have to lift their body weight against gravity while rolling up the curb.

Methods

In this study, the mechanical load was expressed as net joint moments, which is a generally accepted measure to define mechanical load.[11,13,25] Net joint moments are the resulting moments around a joint to compensate for the external moments and to perform a certain task. Therefore, net joint moments are sensitive to the kinematics of the task. If, for the same external load, the kinematics differ, a difference in the net joint moments will be found. Yet no difference will be found if the kinematics do not differ or if the kinematic differences are relatively small compared with the force requirements, as in the weight-relief lift.

In subjects with a high-level SCI, key muscles are often compromised. It was expected that this would become visible in strategy or technique and in external force parameters. However, the subject with a high-level SCI did not perform the tasks in a completely different way; therefore, we did not find differences between the net joint moments among the groups.

The model used in our study was not individualized but based on the morphology of an older cadaver.[23,29] Therefore, we may have under- or overestimated the net moments because the moment component caused by the mass of the limb is constant for different subjects. The choice to use a single model is arbitrary but highlights the effects of kinematics and external forces. Also, our results will be in line with future comparisons between subject groups for individual muscle forces.

The load on the shoulder and elbow is considerable during ADL tasks. Therefore, apart from the ergonomics of the task layout, therapists should be aware of patients' physical capacity before starting to practice these heavy ADL tasks, to prevent early damage to the joints. We believe that overall muscular work capacity plays an important role in the height of the mechanical load. A recent study by Fullerton et al[30] is indicative in this respect; their results showed that highly trained wheelchair athletes experience significantly less shoulder pain than nonathletic wheelchair users. It is expected that for subjects with a high-level SCI, the mechanical load (compression forces, muscle load) can be notably higher—for example, for weight relief—because of the often complete absence of active support from the legs and the compensatory muscle activity in the remaining muscle groups after muscle paralysis. Therefore, an active training or exercise program could be beneficial, probably even in early rehabilitation, to increase work capacity.

The risk for musculoskeletal injuries is not only affected by the peak forces occurring during a task, as presented in this study, but also by the frequency, the duration, the direction of the force, and the point of force application of a given task.[31] Even though the load on the shoulder and the elbows is relatively low during normal wheelchair tasks, wheelchair propulsion is a repetitive task and could lead to overuse injuries as a consequence of the combination of load and repetition. ADL tasks like weight-relief lifting are relatively low-frequent but extremely straining. There is also the absence of sufficient recovery time: subjects must perform lifts during the day and propel themselves. Tasks like making a transfer are performed around 15 times a day,[1] and, as shown by Janssen et al,[32] a physical strain of 60% of the heart rate reserve occurs frequently during transfers. The high loads during ADL tasks might be a risk factor for overuse of the upper-extremity joint, which would be in line with epidemiologic data.[1] When these high loads lead to trauma in the upper extremity, it is likely that no recovery occurs because of the regular (and almost inevitable) repetitive submaximal loading of the upper extremity during wheelchair propulsion. It is therefore likely that neither wheelchair propulsion nor weight-relief lifts by themselves are responsible for the high prevalence of overuse injuries but that the combination of both forms of loading comprise a high-risk factor.

CONCLUSIONS

Negotiating a curb and performing a weight-relief lift were accompanied by a significantly higher net moment in the shoulder and elbow than were found for wheelchair propulsion and reaching. Propelling on a slight slope caused a higher shoulder moment than did normal wheelchair propulsion.

No significant differences were found in the estimated loads on shoulders among the 3 groups. For the subjects with paraplegia, the elbow moments were significantly higher than for the subjects with tetraplegia.

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Suppliers

a. Model 3446; Enraf Nonius BV, Röntgenweg 1, PO Box 810, 2600 AV Delft, The Netherlands.
b. Sunrise Medical Benelux, Pascalbaan 3, 3439 MP Nieuwegein, The Netherlands.
c. Type MC3A-6-1000; Advanced Mechanical Technology Inc, 176 Waltham St, Watertown, MA 02472.
d. Porti, Twente Medical Systems, H. ter Kuilestr 181, 7547 SK Enschede, The Netherlands.
e. Faculty of Human Movement Sciences, Vrije Universiteit, Van der Boechorststr 9, 1081 BT Amsterdam, The Netherlands.
f. Northern Digital, 103 Randall Dr, Waterloo, ON N2V 1C5, Canada.